CLINICAL BIOMECHANICS OF WEAR IN TOTAL HIP ARTHROPLASTY

John J. Callaghan, M.D.; Douglas R. Pedersen, Ph.D.; Richard C. Johnston, M.D.; Thomas D. Brown, Ph.D.

ABSTRACT

Complementary clinical and laboratory studies were performed to identify variables associated with polyethylene wear following total hip replacement, and to elucidate the mechanisms responsible for accelerated wear in the total hip arthroplasty construct.

Observational cohort studies were performed using a prospective clinical database of more than 4000 consecutive primary total hip arthroplasties performed by a single surgeon, to identify wear-related variables. These variables included head size, acetabular/femoral component impingement, and third body debris. Novel digital edge detection techniques were developed and employed to accurately measure wear, and to determine the relationships of head size and third body debris to acceleration of wear. A novel sliding-distance-coupled finite element model was formulated and employed to examine the mechanisms responsible for wear. The long-term cohort studies demonstrated smaller head sizes to be associated with less wear. Third body debris generated from cable fretting was associated with an increase in wear, osteolysis, and acetabular loosening, especially with larger head sizes. The sliding-distance-coupled finite element model replicated the wear rates occurring *in vitro* and *in vivo*, demonstrating the importance of sliding distance on polyethylene wear following total hip arthroplasty. It also demonstrated substantial increases in wear associated with femoral head scratching from third body debris. Further extension of the finite element formulation demonstrated the potential for acetabular component rim damage from impingement wear, and the enhanced potential for third body ingress to the bearing surface with larger head sizes. Edge detection wear measurement techniques demonstrated that early wear rates were predictive of long-term wear rates.

These complementary clinical and laboratory investigations have provided insight into 1) the significance of sliding distance and physiologic loci of motion as contributing factors in minimizing wear, 2) the deleterious effects of third body particulates in accelerating wear, 3) the potential for, and factors related to, impingement wear, and 4) the potential advantages and compromises related to the use of larger head sizes in the bearing surface construct.

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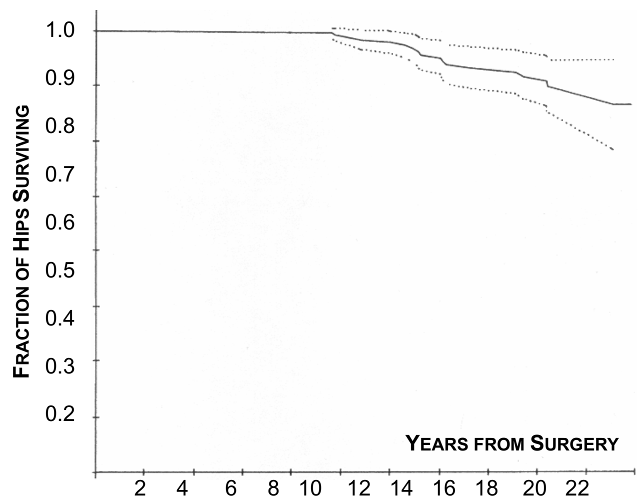


Figure 1. Survivorship curve (solid line) and 95% confidence intervals (dashed lines) for Charnley prostheses implanted by the senior author (RCJ) using first generation cementing techniques. The endpoint is aseptic loosening of the acetabular component, confirmed at revision. (Schulte et al., JBJS 1993).

INTRODUCTION

Almost a decade ago, when the authors reported the first minimum twenty year follow-up of total hip replacement in North America[48], they confirmed the findings

TABLE 1
Results of wear in cohorts of hips with various acetabular-femoral head articulations

Follow-up Period Cohorts	Linear Wear Mean (mm/year)	Linear Wear Standard Deviation	Volumetric Wear Mean (mm^3/year)	Volumetric Wear Standard Deviation
5 Year				
22-mm machined	0.12	0.07	45.22	27.14
22-mm molded	0.11	0.12	40.91	46.97
28-mm molded	0.14	0.13	89.27	79.70
28-mm molded metal back	0.11	0.07	64.70	45.83
7–8 Year				
28-mm cementless metal back	0.11	0.06	65.78	39.01
10 Year				
22-mm machined	0.12	0.06	48.36	24.51
22-mm molded	0.08	0.06	32.71	24.59
28-mm molded	0.12	0.10	70.88	59.61
15 Year				
22-mm machined	0.11	0.07	41.20	25.79
22-mm molded	0.09	0.06	34.59	22.65
20–22 Year				
22-mm machined	0.10	0.07	40.69	26.24

(Pedersen et al., ASTM 1994; Callaghan et al., CORR 1995)

of others that osteolysis, acetabular loosening, and polyethylene wear were the major long term problems associated with the total hip arthroplasty procedure (Figure 1).

Although the conventional wisdom at the time was that head sizes in the range of 28 millimeters were associated with lower wear rates than either 22 or 32 millimeter heads,[28] the authors observed lower linear and volumetric rates of wear with 22 millimeter femoral heads[12] (Table 1). This initial observation, along with our reports[9,30] of osteolysis around secure cementless total hip arthroplasty devices helped redirect investigative efforts away from cement as the leading cause of failure in the total hip arthroplasty construct. Instead, attention moved toward investigation of clinical variables and biomechanical mechanisms associated with polyethylene wear. Since that time, a unique single-surgeon database with well-maintained serial radiographs over a thirty-year follow-up period, and laboratory efforts in experimental and computational biomechanics, enabled the authors to conduct an integrated series of complementary clinical and laboratory studies of total hip arthroplasty wear, aimed at identifying causative factors and elucidating underlying mechanisms.[2,3,7,10,11,12,13,23,25,26,27,29,32,33,34,36,40,44,49,53,54,57]

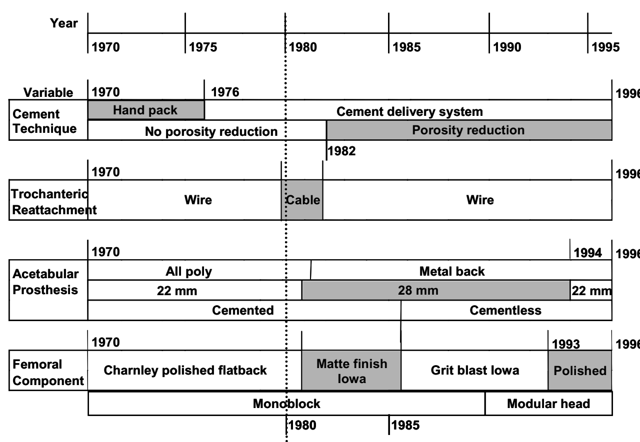
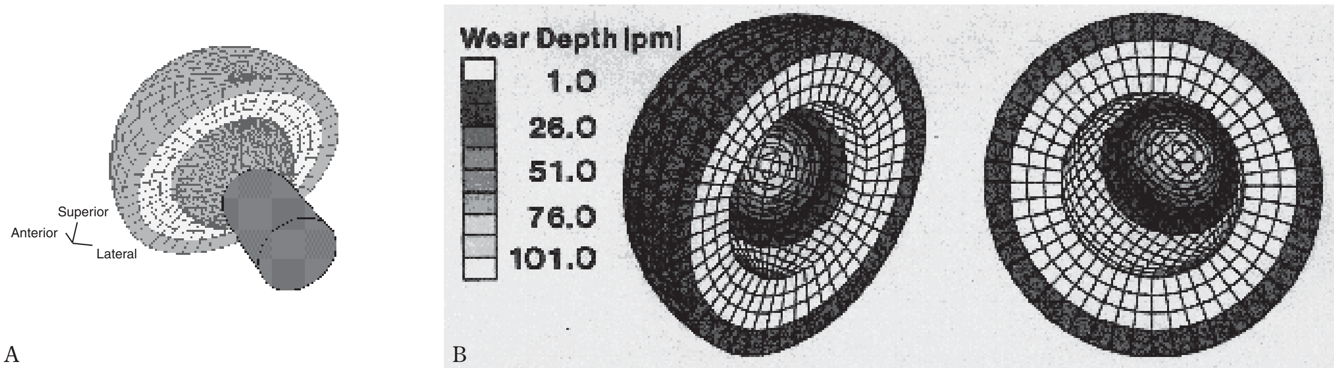


Figure 2. Changes made in prosthesis design or implantation technique over a 26-year period of 4164 hip replacements. Only ten changes were involved, and only two of those changes occurred simultaneously: 22⇒28 mm head size, and Charnley polished flatback⇒Iowa matte finish, in 1981.

The prospective single-surgeon database included more than 4000 consecutive primary total hip replacements. Only a few specific changes were made in either the design or technique, and these few changes



A B

Figure 3. A three-dimensional finite element (FE) model of total hip reconstruction is used to calculate per-gait-cycle wear (in picometers, 10^{-12} meters) of the polyethylene liner of the acetabular component.

occurred at well-prescribed time points. This constituted unique material for performing observational cohort studies related to wear of polyethylene following total hip arthroplasty (Figure 2). Through these clinical studies, the authors identified a number of variables relevant to the understanding of wear mechanisms. This guided the development of laboratory models to elucidate underlying mechanisms of wear.

Over the last decade these authors have tested the following hypotheses:

Hypothesis 1: Across large clinical series, the relative long-term wear performance of contemporary design variants can be reliably predicted directly from their respective articulation dynamics (contact stress and sliding kinematics).

Hypothesis 2: Across large clinical series, long-term polyethylene wear and late loosening can be reliably predicted on the basis of early wear behavior.

Hypothesis 3: In otherwise similar constructs, third body particulate debris causes predictably accelerated polyethylene wear, leading to early radiographic and mechanical failure.

Hypothesis 4: The accelerated effects of third body wear associated with larger head sizes depend on debris access to the bearing surface, mediated by fluid convection.

Hypothesis 5: The polyethylene acetabular component rim damage observed at the time of component retrieval during revision hip surgery is associated with impingement of the femoral head, neck and collar on the acetabular polyethylene liner.

MATERIALS AND METHODS

Clinical Materials

Over a 26-year period the senior investigator (RCJ) performed more than 4000 primary total hip replacements, with very few and discrete changes in the designs and technique used (Figure 2). All radiographs and clinical data were prospectively accumulated, and were used to evaluate revision, radiographic loosening, osteolysis, and wear in a standard manner[4,20] over the last decade.[2,3,7,11,13,23,25,26,29,36,48,54,57] The follow-up studies involved comparisons between cohorts, aimed at identifying how specific changes in implant design or operative technique influenced long term outcome. This group of patients and their radiographs were used to determine variability in wear rates related to femoral head size, to predict late wear rates from early wear rates as determined by digital edge detection techniques, and to evaluate the effect on wear, loosening, and osteolysis of third body debris from cables used to reattach the greater trochanter.

Analytical Methods

To complement the long-term clinical data and to elucidate the mechanisms of wear, as well as to more reliably predict and measure wear, several conceptually new analytical paradigms were developed.

Sliding-distance-coupled finite element model

Three-dimensional nonlinear contact finite element analysis of total hip replacement was linked to interface sliding kinematics, enabling (for the first time) parametric computational study (Figure 3) of polyethylene wear rates and spatial wear distributions.[32,33,34] Hip resultant loads from an inverse Newtonian gait analysis model (validated *in vivo* with an instrumented implant) were used in the FE analysis to determine contact stress distributions on the polyethylene bearing surface (Figure 4) at each of 16 discrete instants of stance phase.[6,16,17,44] Incremental sliding distances of points on the femoral head (Figure 5) were computed from cor-

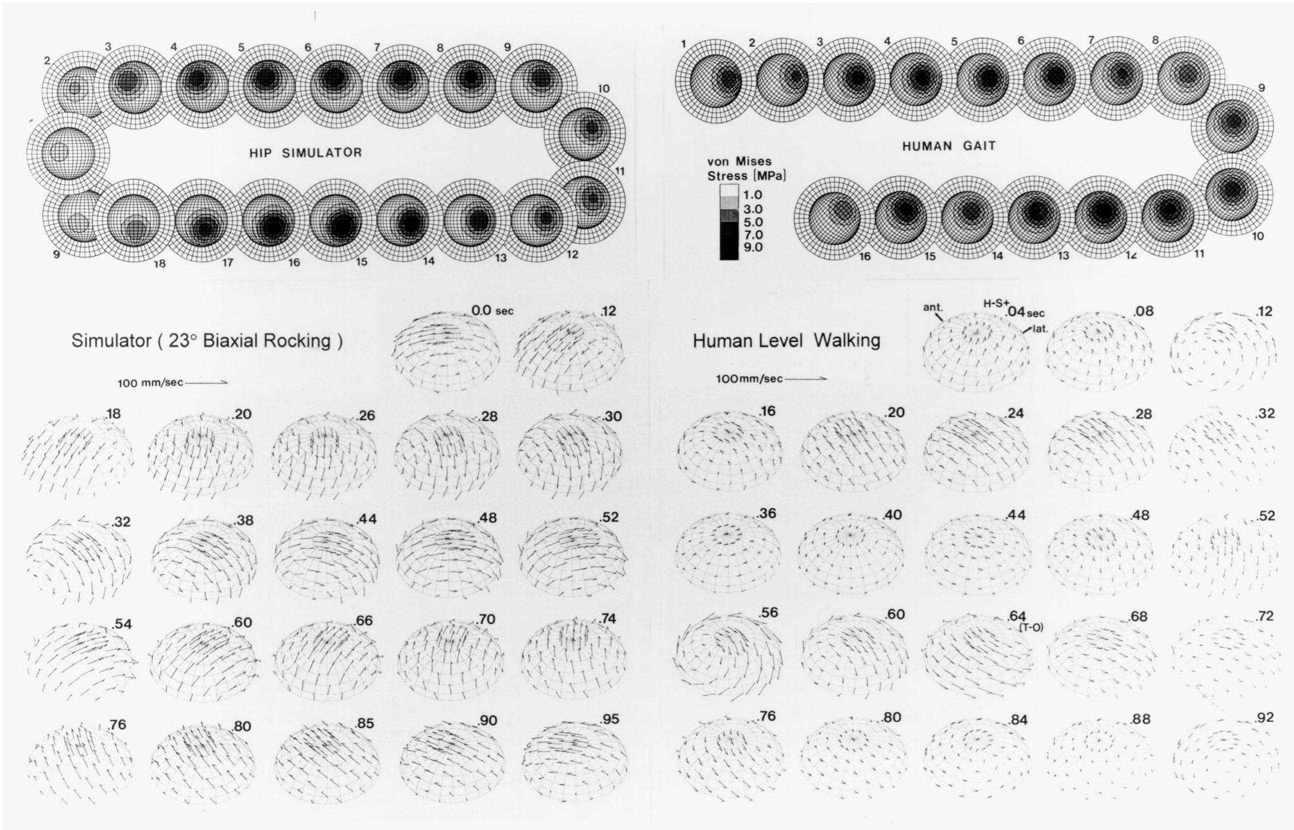


Figure 4. Time-wise variations of contact stress and sliding distance during the articulation cycle in THA. A finite element model (Figure 3a) is used to compute 3-dimensional polyethylene contact stress distributions (contour plots) at serial time points in the gait cycle. Corresponding distributions of bearing surface sliding velocity (vector plots) are determined from recordings of the three-dimensional joint motion patterns. These data are then input to a modified version of the Archard equation. Note that both the contact stress distribution history and the sliding distance history for conventional laboratory wear simulations (left panels) are markedly different from those for human locomotion (right panels), implying very different wear behavior.

responding flexion/extension kinematics.[44] Wear rates were determined by a custom-written computer program that implemented Archard's relationship,[1] coupling contact stress, sliding distance, and a tribologically-based wear coefficient. Later, adaptive remeshing capability was added to the FE model.[34] This feature was introduced to account for conformity changes accompanying progressive removal of polyethylene wear material from the bearing surface, thus allowing for extension of the postoperative wear simulations to the clinically more significant long term regime (as long as 20 years).

The algorithm's temporal convergence (i.e., the minimum frequency of remeshing updates needed to ensure a well-behaved solution) was investigated, with the finding that under most circumstances it was reasonable to make such updates at intervals of about 5×10^5 loading cycles, corresponding to about 6 months of average *in vivo* service. To physically validate the model, a collaborative study was undertaken with colleagues experienced in laboratory wear measurement, to determine whether the finite element model could accurately predict wear occurring in a laboratory hip simulator.[33] The collaborative study design used two cohorts of otherwise identical hemispherical cups, one cohort having a 22-mm bearing surface diameter and the other having a 28-mm diameter. Both cup cohorts were subjected to 3 million cycles at 1 Hz in a biaxial rocking hip simulator, programmed with a Paul-type loading curve, articulating against similarly polished stainless steel balls with bovine serum lubrication. Volumetric wear was measured gravimetrically. Since the true wear coefficient prevailing in the experiments was not known *a priori*, iterative comparisons of computed versus experimentally measured volumetric wear for the 22-mm cups were performed to arrive at a specific wear coefficient value for which computation was brought into identical agreement with the physical measurements. Then, assuming that this same wear coefficient prevailed for the physical testing of the 28-mm cups (all tribological factors

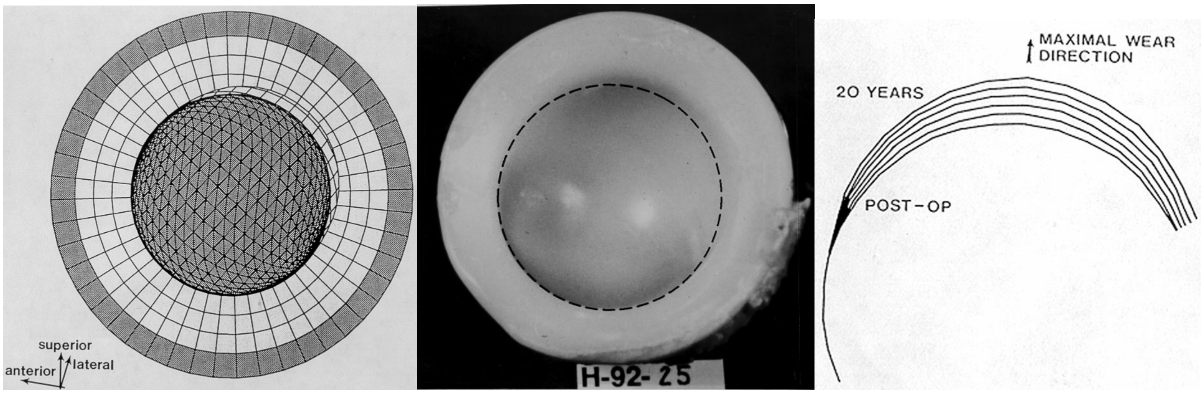


Figure 5. Long-term wear behavior was computed using the adaptively-meshed sliding-distance-coupled finite element model. Material removal at late times was based upon extrapolation of per-gait-cycle wear depth distributions (Figure 3b), but with the finite element mesh (and therefore the contact stress distributions) periodically updated to reflect material removal. Patterns of computed long-term wear (left) were consistent with material loss patterns on retrieval cups (middle), and in profile corresponded to unidirectional "test-tube" wear front advance (right).

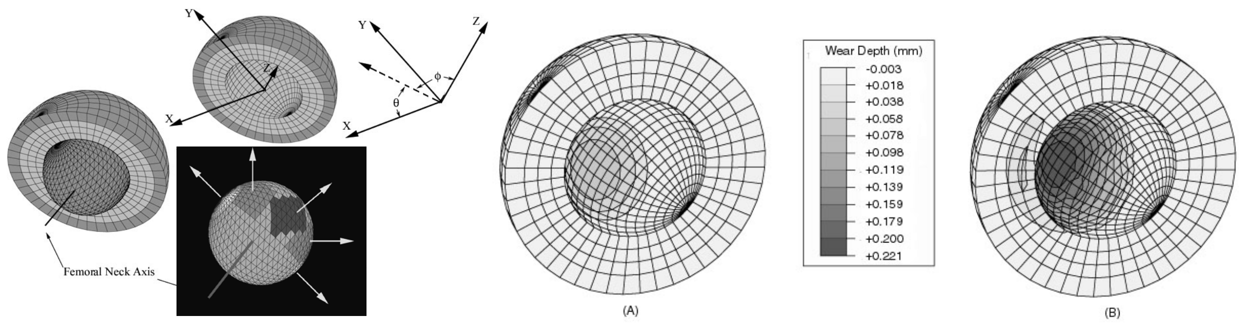


Figure 6. Site-specific wear coefficients are assigned to model local head roughening effects. A patch of Bezier surface facets, which define the femoral head (left panel), could be assigned an elevated wear coefficient value, associated with roughening. Effects of local roughening (wear coefficient = 1.065×10^{-7} mm2/N) on computed acetabular wear were simulated for 10^6 cycles of walking motion (B). Compared to the situation for an undamaged femoral head (A), a 2.13-fold increase in computed volumetric wear was induced. Note also that, for the roughened femoral head, the wear tract becomes less regular than the classic "test tube" pattern.

ostensibly being identical between the two cup cohorts), finite element trials were performed for the 28 mm case, and the volumetric wear results were compared with those obtained physically. Computed wear values agreed with measurements to within 4.1%, a discrepancy of less than one half of one standard deviation of the experimental measurements themselves.

These initial total hip arthroplasty wear simulations involved spatially uniform counterface roughness. These were (gait cycle) temporal-spatial integrations of the product of instantaneous local contact stress σ, times instantaneous local counterface sliding speed ϖ, times a spatially uniform wear coefficient κ, in hip-centered spherical coordinates. To study the effects of head roughening (from head scratches from 3rd body debris) in accelerating wear, algorithmic logic was developed to link head-based and cup-based coordinate systems, thus allowing identification of the head surface site apposing any given acetabular surface site at any given point in time. By means of an automated computational lookup table storing the wear coefficient for each of the large number of sectors (Bezier surface facets) making up the head surface, appropriate time- and site-specific wear coefficient variations were supplied for gait cycle Archard integrations, at each acetabular finite element node. Hence, a non-uniform κ, as in the case of regions of femoral head scratching created by 3rd body wear, could be incorporated at any site of the finite element model femoral head (Figure 6). Even with a relatively small roughened region (5% of head surface area) the resulting volumetric wear was substantially increased, and the direction of wear appreciably altered, relative to computed wear for otherwise similar non-roughened femoral heads.

In addition, the recognition of high levels of wear in acetabular components with rim damage retrieved at revision surgery, and a review of the long-term follow-up of dislocation[8] which demonstrated that 26% of dis-

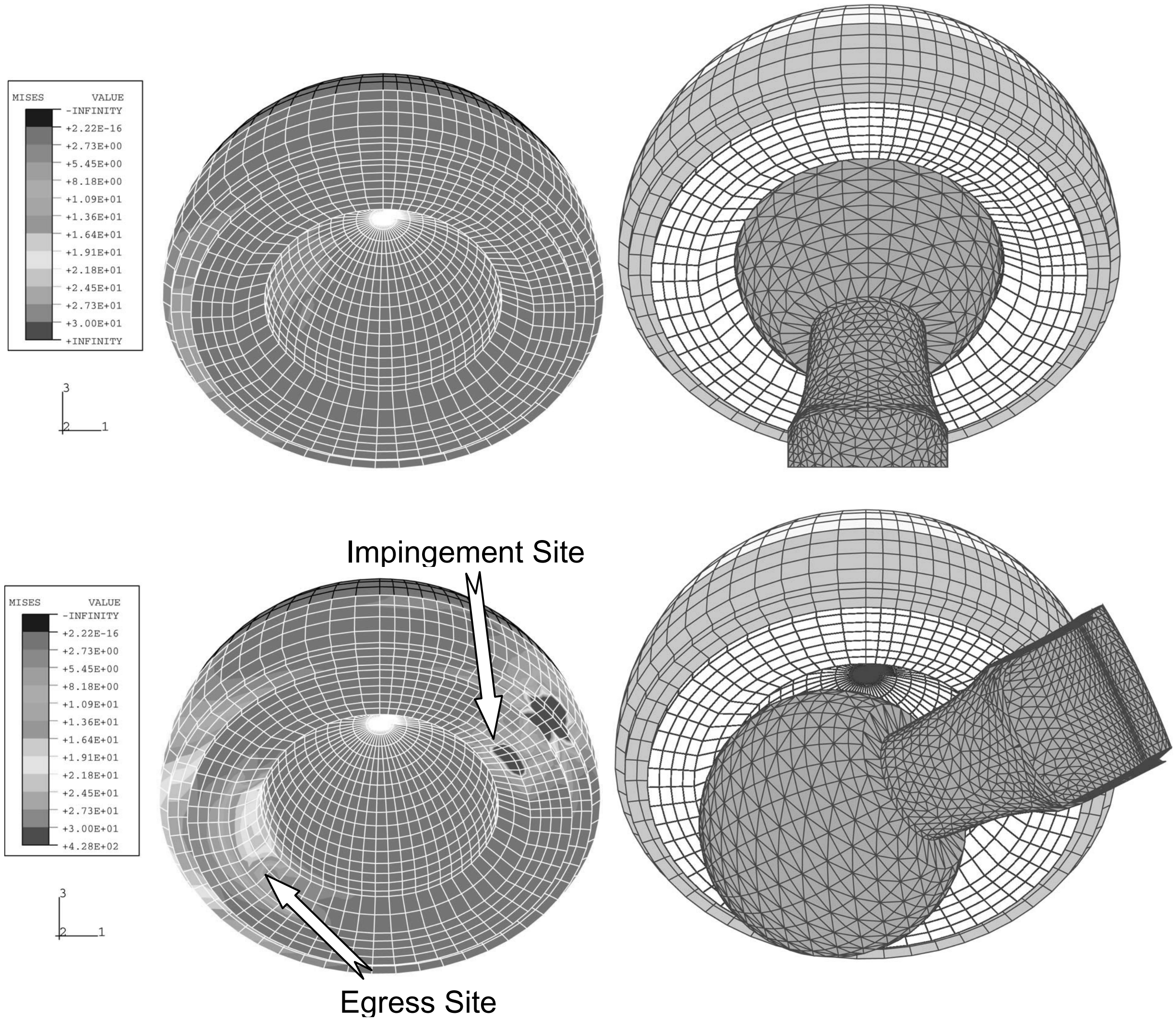


Figure 7. Finite element analysis of the kinetics of total hip impingement, subluxation and dislocation. The model was driven by triaxial motion sequences recorded from subjects undergoing dislocation-prone activities (e.g., leg crossing, rising from toilet seat). The joint loadings were inferred from a 47-muscle inverse dynamics model of the hip.[6] The model articulates normally until neck impingement on the cup. Computed local contact stresses at the impingement and (later) head egress sites are greatly elevated above those for normal articulation, and substantially exceed the yield stress of UHMWPE (lower left panel), even on the outer cup edge, which corresponds with the outer liner wear damage demonstrated in our dislocation studies[8] and in retrieval studies[24].

locations occur late, prompted the authors to use the sliding contact finite element model to study impingement, subluxation and dislocation (Figure 7).

The mechanistic link between rim damage and elevated bearing surface wear is that lever-out subluxation, accompanying impingement, draws debris-laden joint fluid into the opening created. This process was quantified by means of a finite element computational fluid dynamics (CFD) model. The results show that both volumetric fluid ingress and fluid velocities increase with head size (Figure 8), a finding consistent with the clinical observation[23] that larger head size components are preferentially vulnerable to accelerated wear in the presence of a third body challenge. Peak influx velocities for a 28-mm head are 1.29 times those for a 22-mm head, a ratio roughly in proportion to the respective head ratios (1.27), and expected intuitively: For a given distance of head center subluxation, the volumetric rate of fluid influx is proportional to the second power of head radius, whereas the available cross-sectional area available for influx peripherally is proportional to the first power of head radius.

Digital edge detection image analysis wear measurement techniques

Historically, the accuracy and precision of radio-

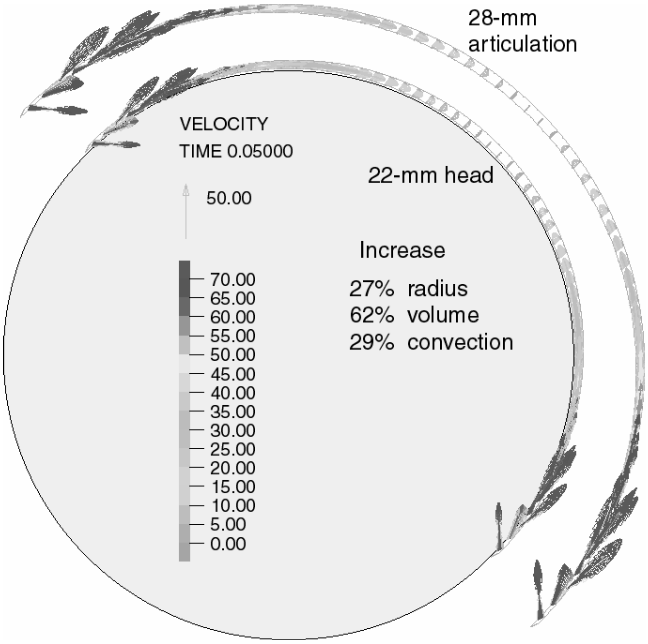


Figure 8. Computational fluid dynamics (CFD) models demonstrate the nonlinear increases in fluid ingress velocities, fluid ingress volume, and concomitant potential third-body debris convection around the subluxating components of increasing head-size total hip constructs.

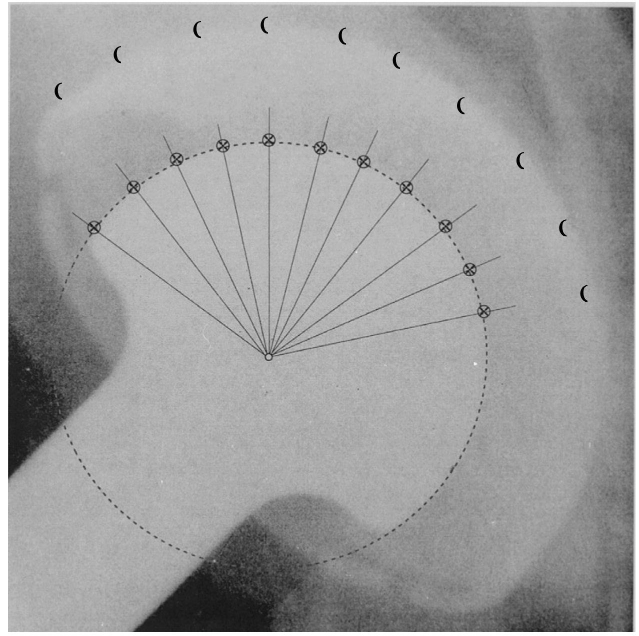


Figure 9. Application of digital edge detection to measure THA wear radiographically. Search rays are computationally generated at 0.5° increments (here, for clarity, rays are displayed only at 10° increments). The pixel grayscale gradient is calculated at each point along each ray. The points of maximal gradient (denoted by the "*f*" symbols for the femoral head and the "o" symbols for the cup backing) identify the respective component margins. Ellipses are least-squares best fit to these two sets of points, to determine the apparent penetration of the femoral head into the acetabular component. Doing this at follow-up, and subtracting the corresponding measurement postoperatively, allowed assessment of interval wear between those two time points.[51,52,53]

graphic measurements of penetration of the femoral head into the acetabular component had been compromised by reliance upon subjective–manual assessment of head penetration.[14,15,18,28,37,45,55,56] To reduce the degree of subjectivity entering into head penetration measurements, digital edge detection image analysis was introduced to more accurately determine the amount of head penetration.[51, 52] The edge detection technique allowed the first-ever automatic, fully objective penetration measurements, using ellipses best-fit to hundreds of component surface points that were identified computationally as sites of maximal local gray-scale gradient (Figures 9 and 10). The new penetration measurement technique was validated by measuring wear artificially produced by spherical-front milling of polyethylene liners in bench top series. Under such conditions, digital edge detection proved 6.4 times more accurate, and 7.1 times more reproducible, than manual measurements made with conventional circular templates.[53]

RESULTS

Wear measurements using a circular template technique were performed on consecutive series cohorts of patients with minimum five-year follow-up radiographs (210 hips). Significantly less wear was demonstrated for 22-millimeter head versus 28-millimeter head components[12] (Table 1). There was a strikingly wide range in the wear rates within the individual cohorts (0 to 0.8 mm per year). In addition, the wear rate distributions were strongly non-Gaussian, skewed by the small number of outlier patients with very high wear rates.[41] Digital edge detection techniques demonstrated lower wear rates with cementless acetabular components with 22-millimeter head sizes than with 28-millimeter head sizes (Figure 11).[43]

The sliding-distance-coupled finite element model was able to reproduce the wear patterns occurring during gait versus in biaxial rocking wear simulators (Figure 4). The model also accurately demonstrated the volumetric wear increase with increases in head size (Figure 12).[32, 34]

To evaluate the ability to determine long-term polyethylene wear from early femoral head penetration (at two years), edge detection techniques were utilized on 197 consecutive total hip replacements with 1,237 archived radiographs taken over a 10 year follow-up period (Figures 13 and 14). The digital edge detection measurements were analyzed using a novel random-coefficients statistical formulation, developed specifically

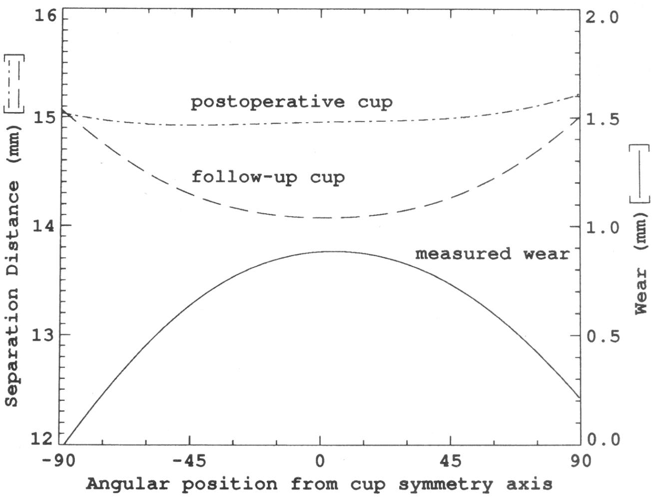


Figure 10. Distribution of the separation distances between ellipses on consecutive radiographs (left axis). Point-by-point subtraction of the distribution at the latest follow-up evaluation from the postoperative distribution represents polyethylene wear (right axis). This also allowed precise determination of the direction of maximum linear wear (i.e., peak of the measured wear curve).

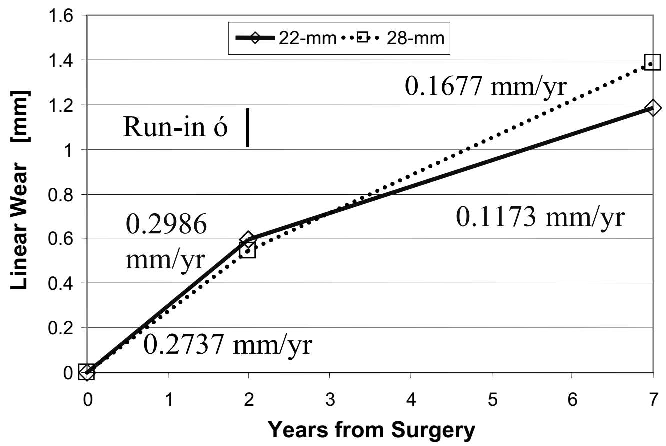


Figure 11. Group-average behavior for "bedding-in" (run-in) and long-term penetration into the polyethylene cups, by 22-mm and 28-mm femoral heads.

to address the distribution of wear rates observed across this population across time. A formal regression equation was reported, by means of which the ten-year wear depth could be quantitatively estimated from a given patient radiograph at the two-year follow-up visit. Series-wide, the correlation between predicted versus observed late wear depths was r = 0.683. Since many very early (less than 2 year) radiographs were available, the authors also used this database to make the first quantitative observations of the initial clinical "bedding-in" process, behavior which could be described mathematically in terms of a best-fit decaying exponential function. Finally, differences in wear direction were detected between hips, and also between subsequent time intervals for the same hip.[40]

Evidence for accelerated polyethylene wear due to third body debris was studied in 709 consecutive primary total hip arthroplasties performed over a five-year period overlapping a thirty-month interval of braided cable usage (Figure 15), all of which patients were followed for a minimum of ten years. The sequential switch from wire to braided cable coincidentally occurred during a period of transition from use of a Charnley femoral component with an all-polyethylene acetabular component, to an Iowa femoral component with an all-polyethylene acetabular component, and then to an Iowa femoral component with a metal-backed acetabular component. Thus, accelerated wear due to elevated third-body debris burden could be confirmed for three distinct implant constructs. Across the entire study population, (fretting prone) braided cable reattachment hips had significantly more acetabular wear ($p<.05$), osteolysis ($p<.0001$), and radiographically apparent loosening ($p<.001$) than did the wire reattachment group. The effect of cable debris on wear was less pronounced with 22-mm head articulations,[23] a finding consistent with the CFD finite element model.

Using the sliding distance-coupled finite element model to simulate femoral head roughening regions of various sizes, severities, and locations resulted in appreciable changes in the computed acetabular wear patterns, including shifts in vectorial wear direction, distortions of wear front sphericity, and changes in volumetric wear patterns (Figure 6). These findings reflect the variability in wear direction noted with edge detection radiographic measurements, and seen in retrieved acetabular components.[5,19,21,22,24,35,46,47,58,59] The fluid mechanics model demonstrated the fluid pressures associated with particle convection into the articulating surface of a larger head construct, a plausible explanation for the accelerated wear observed with 28-mm versus 22-mm head constructs in the cases with cable debris.

The computed local contact stresses (39 MPa) at the impingement and head egress sites in the impingement/dislocation model substantially exceeded the yield stresses of ultrahigh molecular weight polyethylene (Figure 7). Larger femoral heads with a constant head/neck ratio demonstrated a decrease in these impingement stresses, and in the corresponding head egress stresses. The dislocation FE model also demonstrated the gain in stability achievable with increase in head size: increases of approximately 0.59 N-m of peak resisting moment per millimeter of head diameter increase, up to 44 mm.

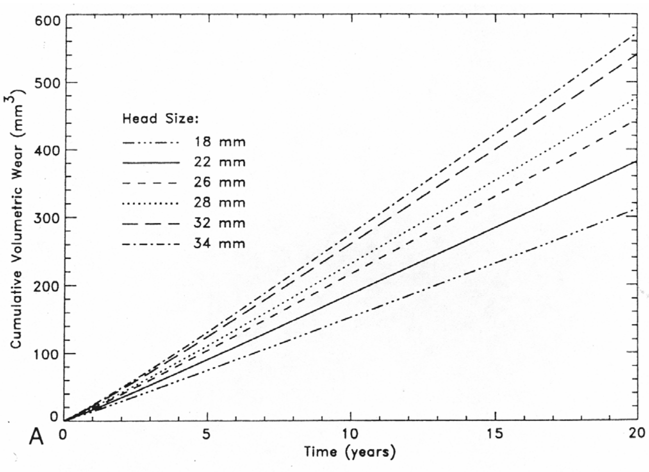


Figure 12. Effect of the head size on time-dependent accrual of volumetric wear, up to as many as 20 years. Volumetric wear increased in proportion to increases in head size.

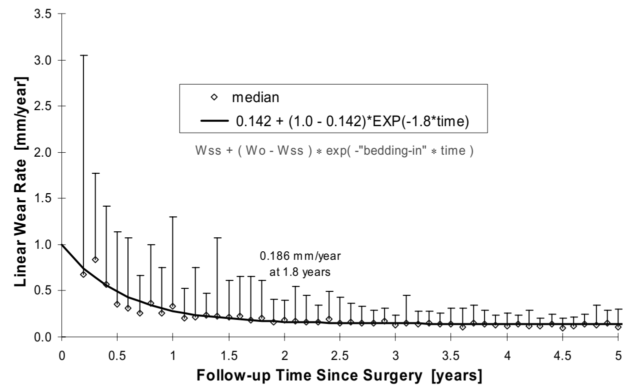


Figure 14. The initial clinical "bedding-in" process could be described in terms of a decaying exponential (best-fit), to quantify early-term femoral head behavior. The time point to achieve steady state wear (Wss), at which 95% of the initial transient concluded, was 1.8 years. (This figure has subsequently been rounded to 2 years to simplify description of the "bedding-in" period.[39])

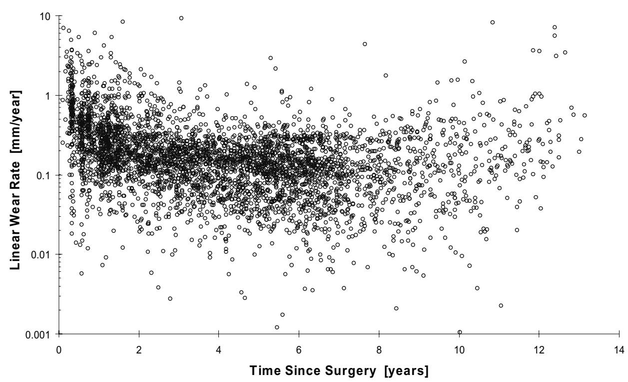


Figure 13. Temporal evaluation of radiographically apparent interval wear rates. Data for a single patient cohort are illustrated as a scattergram of individual interval linear-wear rate measurements (197 hips, 1237 archived radiographs taken over a 14-year follow-up period).[40]

DISCUSSION

This paper summarizes a ten-year period of clinical and laboratory investigation of wear associated with the long-term performance of the total hip arthroplasty construct. Using complementary clinical and laboratory studies (enabled by the unique 30 years clinical and radiographic database, and by novel laboratory computational and experimental techniques), the following questions were addressed:

1. Can long-term wear performance be reliably predicted directly from articulation kinetics (contact stress and sliding distance)?
2. Can long-term wear be reliably predicted on the basis of early wear behavior?
3. Do third body particulates cause accelerated polyethylene wear and early total hip arthroplasty construct failure?
4. What mechanisms are responsible for acetabular component polyethylene rim damage and wear?

This body of work included the first, or among the first, reports that, in the clinical situation, the finding that wear rates with conventional polyethylene are lowest with 22 millimeter femoral heads.[11,42,48] This was later corroborated with autopsy and revision retrieval studies.[24] The sliding-distance-coupled finite element model, initially developed in 1995,[31] documented for the first time that favorable articulation kinematics were the key reason for the lower wear rates observed with 22-millimeter heads, rather than head-size-dependent wear being attributable to Charnley's concept of low friction.[15] The digital edge detection techniques, introduced for the first time in 1994[51,52] opened the way to more accurately determine wear, and demonstrated that accurately measured early wear rates predicted long term wear. This body of work also included among the first, if not the first, quantitation of the bedding-in period associated with wear, and highlighted the need for advanced statistical means for analyzing wear, due to the non-Gaussian distribution of wear rates within cohorts.[38,39,40]

The unique opportunity to study an unfortunate group of hip replacement patients who encountered elevated third body debris due to fretting of trochanteric cable, enabled documentation, with minimum 10-year follow-up, of the accelerated rates of polyethylene wear, osteolysis and component loosening attributable to third body debris migration to the bearing surface.[23,27] Subsequent use of the finite element sliding-distance-coupled formulation to study femoral head surface roughness changes, consistent with third-body-gener-

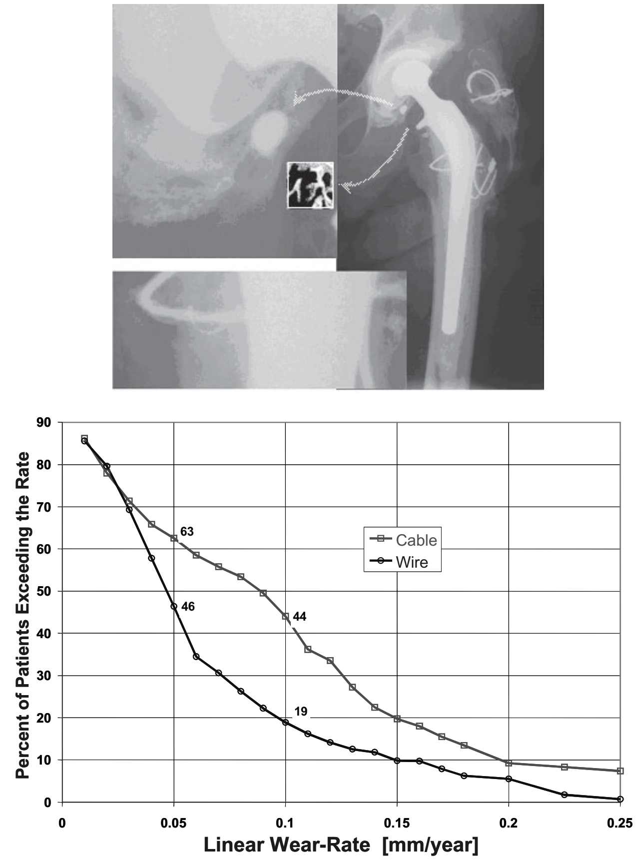


Figure 15. Effects of elevated third body debris burden on in vivo wear. Two patient cohorts were compared, both with the same implant: an Iowa femoral component articulating with a cemented 28 mm metal-backed Tibac (Zimmer) cup. One group ("Cable," 197 consecutive patients) had trochanteric fixation with 1.5 mm, 7-strand Co-Cr-W-Ni cable, subsequently found to be fretting prone (top) and its usage discontinued. The other group ("Wire," 157 consecutive patients) had trochanteric fixation with single-strand stainless steel wire. Series wide (bottom) there is a much larger fraction of problem "Cable" patients (10%) with linear wear rates exceeding the clinically problematic rate of 0.2 mm/year.

ated scratches, provided a plausible explanation for the extreme variability of wear vector direction demonstrated in autopsy acetabular component retrievals, and in serial digital edge detection wear measurement studies.[40,59] The computational fluid mechanics model demonstrated that pronounced fluid convection differences between smaller and larger femoral head articulations are a plausible explanation for the less deleterious effects on bearing surface wear for cable debris cases with 22-mm head articulations than with 28-mm head articulations.

Finally, the application of the finite element formulation to study component impingement and dislocation, the first of its kind, has helped explain the association between impingement-related acetabular component rim wear and dislocation following total hip arthroplasty.[49,50]

In the future, this same general approach of using complementary clinical and laboratory studies should prove useful to characterize and evaluate the wear mechanisms encountered with the new highly-crosslinked polyethylenes and hard bearing surfaces, and issues such as the benefits and compromises associated with the use of larger head sizes (i.e., 36 to 44 millimeters) in conjunction with these new bearing surfaces. In addition, continued work to elucidate the mechanisms by which third body debris enters the bearing surface of total hip replacement constructs should aid the evolution of component designs and surgical techniques to reduce this major additional risk to construct longevity.

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